A method and an apparatus for processing auscultation
signals.

This invention relates to a method of processing auscul-
5   tation signals, comprising the steps of receiving an aus-
cultation signal and providing a biased auscultation sig-
nal, and calculating a signal representative of an esti-
mated rhythm of the auscultation signal.

Moreover, the present invention relates to an apparatus
10   for processing auscultation signals, and in particular to
an electronic stethoscope suitable for use in cardiology.

In recent years physicians have had an impressive arsenal
of instrumentation at their disposal for the diagnosis of
cardiovascular diseases. One such instrument is the well
15   known stethoscope used to detect sounds originating from
the heart and adjacent large vessels. With the introduc-
tion of the stethoscope, auscultation grew into an art
providing the clinician with valuable information of the
functional integrity of the heart. Nowadays it is possi-
20   ble to process the information residing in the ausculta-
tion signal electronically by use of embedded knowledge
obtained by clinical research. Therefore it is possible
to relieve the clinician from trivial tasks. One such
task may be to estimate the heart rate of a beating heart
25   based on the auscultation signal.

The auscultation signal comprises characteristic sounds
that originate from the contraction and expansion of the
heart in the process of pumping blood, i.e. sounds caused
by vibrations from acceleration and deceleration of
30   blood. When a beating heart is auscultated it is not the
beat frequency that is heard, but rather the characteris-
tic sounds. Further, for mid-aged healthy persons in nor-

mal conditions, the characteristic sounds often denoted S1 and S2 related to a contraction and expansion of the heart, respectively, are separated in time by a sufficiently large interval that allows for perception of the characteristic sounds as individual sounds, i.e. the characteristic sounds, are typically separated by more than 150-200 milliseconds for adults and about 50 milliseconds for children. Typically, the characteristic sounds have a continuous frequency spectrum distributed between approximately 15 Hz and 100 Hz and peaking at 30-40Hz.

However, the characteristic sounds may be superimposed with sounds originating from different types of vibrations in the heart. These sounds are the so-called murmurs, that may be caused by mechanisms such as blood turbulence or flutter of structures. Murmurs may be divided into two primary groups: a first group comprise random 'noisy' sounds and a second group comprise 'musical' sounds with a continuous narrow-band frequency spectrum. Experience has shown that in a human being the 'musical' sounds are typically related to the heart beat frequency. The frequency spectrum of murmurs may be distributed up to 500-2000Hz.

The characteristic sounds may also be disturbed by spurious contractions of the heart appearing at certain or random points in time. Further, the characteristic sounds may also appear at random points in time. This phenomenon is known as arrhythmia.

When the sound of a beating heart including murmurs, if any, is auscultated by means of a transducer placed on e.g. the chest of a patient, then low-frequent rumble noise will be picked up by the transducer. This rumble noise may have a frequency content primarily distributed

from just below the frequency of the heart rate towards DC.

In severe cases these disturbances are able to destroy the characteristic sounds in the auscultation signal com-
5   pletely or at least disturb the auscultation signal to such a degree that it may be impossible to estimate even an approximated heart rate by use of the prior art tech-nology.

However, for a trained person using a non-electronic
10   stethoscope, the information in the form of characteris-tic heart sounds is relatively easily recognisable from repetition to repetition and therefore relatively easy to count. This is due to the fact that the human perception of the auscultation signal is very good at diminishing
15   the disturbances or artefacts and pay attention to the features in the auscultation signal that provides infor-mation for estimating the heart rate.

US patent 4,972,841 discloses an electronic stethoscope with a pulse rate display for displaying an estimated
20   pulse rate. In the stethoscope a transducer converts so-called Korotkoff sounds into electrical signals. Korotkoff sounds are sounds created by vibrating walls of collapsed vessels as blood is just allowed to flow through the vessel. The electrical signals are amplified
25   and fed to a counter in which the detected pulse rate per unit time is calculated and the result is displayed as a digital pulse rate.

However, this patent is not concerned with estimation of heart rates originating from humans where the measured
30   sound may be very different from expected normal heart sounds. The patent does not disclose any means adapted to suppress artefacts that may be confused with or mixed up

with the information in the auscultation signal relating to the heart rate.

US patent 4,436,096 discloses a digital heart rate stethoscope. A transducer produces electrical signals representing the heart sound of a human patient and a filter is connected to receive the electrical signals. The filter is a narrow-band band-pass filter whose passband is centered on a characteristic heart sound frequency e.g. 33 Hz. The filter is thereby adapted to improve the signal-to-noise ratio and enables the transducer to be used over a patient's clothing. The filtered signal is converted into pulses which are counted in a counter and displayed as a digital heart rate indication.

Again, this patent is not concerned with estimation of heart rates originating from humans where the measured sound may be very different from the expected normal heart sounds. Neither does this patent disclose any means adapted to suppress artefacts that may be confused with or mixed up with the information in the auscultation signal relating to the heart rate.

International Application WO 97/00045 discloses a stethoscope for examining the characteristic of sound waves made by a beating heart. The stethoscope comprises means for establishing a threshold wave form beyond which cyclic reference complexes appear and are identified, and a time measurement device for measuring the time elapsed between similar moments on adjacent reference complexes as an indicator of rate of heart beat.

In this stethoscope the heart rate is estimated based on the assumption that there is substantially no noise between the heart sounds originating from contraction and expansion of the heart. The stethoscope is therefore not

suited for use of estimating heart rates of persons suf-
fering from heart murmur or arrhythmia.

Other apparatuses for estimating the heart rate use means
for determining the envelope of the auscultation signal.
5   The envelope is determined by taking the absolute value
of the auscultation signal and band-pass filtering the
absolute value signal.

Consequently, the prior art involves the problem that the
heart rate is estimated erroneously when some degree of
10   murmurs and/or arrhythmia is present in the auscultation
signal.

Thus the primary object of the invention is to provide a
more precise and robust heart rate estimation, estimating
the heart rate that originates from a real and blood
15   pumping contraction and expansion of the heart.

This is achieved, when the apparatus mentioned in the
opening paragraph is characterized in that the estimator
is adapted for selecting at least a part of the biased
auscultation signal as first signal and calculating the
20   conformity between the first signal and the biased aus-
cultation signal.

Consequently, it is possible to suppress artefacts and
random noise in the auscultation signal that may be con-
fused or mixed up with the information relating to the
25   heart rate. There is thereby obtained a signal in which
the heart rate is very well defined.

In an expedient apparatus for processing auscultation
signals, the estimator calculates a cross-correlation
function. This cross-correlation function supplies a
30   well-defined signal having large signal values at good
conformity and small signal values at poor conformity.

When the part of the biased auscultation signal repre-
sents one of a succession of cycles of the biased auscul-
tation signal, the part may represent a reference cycle
that is found ideal in some aspects.

5 In another expedient apparatus for processing ausculta-
tion signals, the estimator calculates an auto-
correlation function. The auto-correlation function is
especially expedient when the heart rate is relatively
stable over a given time interval.

10 The quality of the auscultation signal can be validated
by verifying at least one of the following three items in
a signal representing the conformity of the auscultation
signal:

a) time differences between located extreme values must
15 be within predetermined limits;

b) minimum and maximum time differences in proportion to
the mean of the time differences must be within prede-
termined limits;

c) the magnitude of the result of the correlation at the
20 extreme values location must be within predetermined
limits.

Then a preferred quality level may be defined by means of
the preferred limits that are related directly to the
statistics of rhythm estimation.

25 The bias processor may comprise a filter for calculating
an A-weighted version of the auscultation signal or an
approximated A-weighted version of the auscultation sig-
nal. Simulations have shown that an A-weighted version of
the auscultation signal comprises sufficient information
30 to estimate the heart rate in a consistent way in a vari-
ety of situations.

When the A-weighted version of the auscultation signal is calculated by means of an approximation corresponding to a double differentiation of the auscultation signal in a predetermined frequency range, the filtering process re-
5   quires a minimum of computational effort.

The bias processor may comprise an adaptive band-pass filter for filtering the signals provided by the envelope detector; said band-pass filter at least having an upper and a lower pass-band, respectively, to select from; said
10   adaptive band-pass filter comprising a controller select-ing the lower pass-band when a relatively large fraction of a signal input to the band-pass filter is low-frequent, and selecting the upper pass-band when a rela-tively low fraction of a signal input to the band-pass
15   filter is low-frequent. Thus, it is possible to estimate the rhythm in the auscultation signal over a relatively wide range of frequencies while it is avoided that the signal-to-noise ratio is ruined, i.e. it is possible pre-serve a relatively narrow pass-band of the band-pass fil-
20   ter.

The auscultation signal may comprise samples that arrive at a sample rate, and the apparatus may comprise a syn-chronous processor operating at a rate corresponding to the sample rate, and further comprising an asynchronous
25   processor operating at time intervals that are initiated by a request. Hereby, it is possible to acquire new sam-ples of the auscultation signal in real time while a cor-relation function operating on prior sample values of the auscultation signal is calculated.

30   Alternatively, the object of the invention is achieved when the apparatus is characterized in that the bias processor comprises a filter having a frequency response

corresponding to an A-weighing or an approximated A-weighing, at least for a frequency range of interest.

Consequently, it is possible to estimate the heart rate of a beating heart, based on the characteristic sounds of the beating heart and murmurs related to the heart rate. Thus, the heart rate may be estimated based on murmurs relating to the heart rate if the characteristic sounds do not provide sufficient information for estimating the heart rate. Further, low frequency rumble noise is diminished and the high frequency components are enhanced, thereby providing well defined information of the relative location of the characteristic sounds with respect to time. This well defined information is suitable for estimating the heart rate electronically.

When the auscultation signal is filtered with a filter having a frequency response corresponding to a double differentiation, the computational effort required to calculate the A-weighted version of the auscultation is reduced.

The frequency range of interest may span the frequency range below 2000Hz. The characteristic heart sounds and murmurs related to the heart rate comprise frequency components located in this frequency range and provide sufficient information in this range for estimating the rhythm of a beating heart.

Moreover, the invention also relates to methods of estimating the rhythm in auscultation signals.

The invention will be explained more fully below in connection with a preferred embodiment and with reference to the drawing, in which:

fig. 1 shows a block diagram of a processor for processing auscultation signals by means of a bias filter calculating an approximated A-weighted auscultation signal;

fig. 2 shows the expected frequency spectrum of the characteristic heart sounds, a murmur, and rumble·noise;

fig. 3 shows a block diagram of a processor for calculating the beat frequency of an auscultation signal by means of a correlator;

fig. 4 shows a block diagram of a processor for calculating the beat frequency of an auscultation signal by means of a correlator and a bias filter comprising A-weighing;

fig. 5 shows an adaptive band-pass filter;

fig. 6 shows signals related to an embodiment using a correlator and a bias filter comprising A-weighing;

fig. 7 shows an example of the information in the envelope of the auscultation signal being insufficient;

fig. 8 shows an example of the information in the envelope of the A-weighted auscultation signal being insufficient.

Fig. 1 shows a block diagram of a processor for processing auscultation signals by means of a bias filter calculating an approximated A-weighted auscultation signal. An auscultation signal is a signal representing the acoustical signal that may be picked up by some type of transducer located to provide an auscultation signal representing the sound of a beating heart. The auscultation signal is hereinafter denoted VCG and may be a digital or analogue signal.

The block diagram comprises a bias processor 106. The VCG signal is filtered in the bias filter 101. The frequency response of the filter corresponds to an approximated A-weighing of the VCG signal. A-weighing is a linear model of the human ear modelling the human perception of signal amplitudes in the audio frequency range, i.e. about 20-20000Hz. A-weighing is typically modelled as a second order band-pass filter with a centre frequency of 2000Hz. The A-weighing model is approximated as a double differentiation of the VCG signal with respect to time thereby providing an A-weighted version, A-VCG, of the VCG signal. The approximation is valid for frequencies between about 20-200Hz. The bias filter 101 may thus be viewed as a second order high-pass filter slope. The filter thereby diminishes low frequency rumble noise that may be present in the auscultation signal. Further, the filter enhances the upper part of the frequency spectrum of the characteristic heart sounds and eventually murmurs. However, the A-weighted signal A-VCG may also enhance noise originating from sources that are not related to either the characteristic heart sounds or eventually murmurs thereby destroying the signal-to-noise-ratio. The A-weighted signal A-VCG is therefore low-pass filtered by means of the filter 102. The low-pass filtered signal is then supplied as an input signal to the rectifier 102 that calculates the absolute value of the input signal and thereby provides an output signal. The output signal from the rectifier 103 is band-pass filtered in the filter 104. This band-pass filtered signal thereby constitutes the envelope of the A-weighted signal A-VCG. The band-pass filter has a pass-band corresponding to the range of expected heart rates e.g. from 0.5 to 3.3HZ. Further, the stopband of the band-pass filter diminishes any DC component and frequency components above the fundamental heart beat frequency, i.e. relative high frequency components.

The information in the band-pass filtered signal present as repetitive well-defined cycles is thereby representative of the heart rate. The heart rate may be estimated by means of an estimator 105. The estimator may be in the

5 form of a peak locater determining the timing of the well-defined extreme values of the band-pass filtered signal. Alternatively, the estimator may be in the form of a threshold or zero-crossing detector thereby generating a square signal with a frequency representing the es-

10 timated heart rate. When the estimator comprises a threshold detector, the biased auscultation signal and a threshold value are both input to a comparator. The threshold value may be in the form of a constant value or an adaptive threshold value. The adaptive threshold value

15 may be provided by a low-pass filter filtering the biased auscultation signal.

Fig. 2 shows the expected frequency spectrum of the characteristic heart sounds, a murmur, and rumble noise. The spectrums are rough approximations and for illustration

20 purposes only. The magnitude Amp. of the spectrums are in arbitrary units and shown as a function of the frequency F [Hz]. The frequency spectrum 301 of the characteristic heart sounds related to a contraction and an expansion of a beating heart is illustrated roughly. In the literature

25 the sound related to a contraction is denoted S1 and the sound related to an expansion is denoted S2. The frequency spectrum of the frequency contents of S1 and S2 is primarily located between 20 and 50Hz but may be distributed up to 100-200Hz and down to below 10Hz.

30 However, the frequency contents of murmurs are typically distributed at higher frequencies. This is shown by the frequency response 302. The frequency contents of murmurs comprising 'musical' sounds is distributed over a relatively narrow-band frequency range 306. This relative

narrow band comprises information that may be used for estimation of the heart rate. Typically, murmurs are also associated with a relative broad band 205 not comprising information regarding the heart rate. An absolute numeri-
5 cal definition of these frequency bands and the magnitudes of the auscultation signal in these bands will depend on the murmur in question.

The frequency spectrum 304 of low-frequent rumble noise typically overlaps the frequency spectrum 301 of the
10 characteristic heart sounds.

The transfer function 303 shows a preferred transfer function for a filter for calculating a biased auscultation signal. The filter is thereby able to enhance the amplitude of the frequency components of the upper part
15 of the frequency spectrum of the characteristic heart sounds and the amplitude of the frequency components of the frequency spectrum of murmurs. Further, low-frequency rumble noise that may be present in the auscultation signal is diminished. The transfer function 303 comprises a
20 high-pass 40 dB per decade slope. The slope may be a part of an approximated A-weighing function, where the approximation is implemented as a second order band-pass filter with a centre frequency of 2000Hz.

Fig. 3 shows a block diagram of a processor for calculat-
25 ing the beat frequency of an auscultation signal by means of a correlator. In this embodiment the received auscultation signal VCG is a digital signal comprising digital samples arriving at a sample frequency fs, e.g. fs equal to 4000Hz.

30 The embodiment comprises a bias processor 201 processing the VCG samples at a rate synchronous with the sample frequency and an estimator 302 capable of operating asynchronous relative to the bias processor and providing a

signal HR representative of the heart rate, when it is requested. The estimator 302 receives data in the form of samples from the bias processor 301 via the buffer 303. The buffer 303 may be in the form of a circular buffer or
5   a first-in-first-out (FIFO) memory.

In the bias processor 301 the digital auscultation signal VCG is received and the respective absolute values of the VCG signal are calculated by means of the rectifier 304. Then the absolute values are band-pass filtered by means
10   of the filter 305 in order to calculate the envelope of the auscultation signal. The band-pass filter has a pass-band corresponding to the range of expected heart rates e.g. from 0.5 to 3.3HZ. Further, the stop-band of the band-pass filter diminishes any DC component and fre-
15   quency components above the fundamental heart beat fre-quency, i.e. relatively high frequency components.

Samples from the band-pass filter are supplied to the buffer 303 at even points in time. The buffers may e.g. be circular buffers that have a capacity for a number of
20   sample points, e.g corresponding to a signal sequence of 5 seconds. The oldest sample in the buffer is overwritten when a new sample arrives from the band-pass filter. Sam-ples are taken from the buffer when the estimator 302 is initiated. The estimator may be initiated at certain
25   regular or irregular points in time e.g. in order to pro-vide information for the update of a digital display.

Although the envelope signal provided by the band-pass filter via the buffer 303 is smooth and comprises repeti-tive cycles it may be difficult to determine whether one
30   or more cycles are present during a specific time inter-val. However, the information content in the envelope signal may be enhanced towards determining the exact num-ber of cycles in a specific time interval. This is done

by calculating the conformity of the envelope signal with respect to the envelope signal itself. This is based on the assumption that the vast majority of the cycles in the envelope signal represents an actual interrelated contraction and expansion of a heart in the process of pumping blood. The conformity of the envelope signal is calculated by means of the correlator 306 calculating the auto-correlation of the envelope signal.

In a preferred embodiment the correlator comprises means for selecting that cycle in the envelope signal that has the largest amplitude. This is done under the fair assumption that the largest amplitude in the auscultation signal is caused by a normal contraction/expansion of the heart that is least likely to comprise any artefacts. The extent of the selected cycle is found by searching for the nearest local minima that is less than zero. Thereby, falling into local minima with a magnitude greater than zero is avoided. This search for a minima is applied both forward and backward relative to the position of the largest amplitude. The selected cycle is stored and cross-correlated with the rest of the envelope signal. Alternatively, convolution or any other function capable calculating the conformity of a signal with respect to the signal itself may be used.

Fig. 4 shows a block diagram of a processor for calculating the beat frequency of an auscultation signal by means of a correlator and a bias filter comprising A-weighing. In this preferred embodiment the bias filter comprising A-weighing and the correlator are combined in order to obtain an enhanced overall performance, i.e. it is possible to estimate the rhythm of a beating heart even though the characteristic sounds of the heart are distorted severely.

The auscultation signal VCG is in the form of a digital signal comprising sample values arriving at a sample frequency fs, e.g. fs = 4000Hz. The auscultation signal is branched. A first branch supplies the auscultation signal to a rectifier or an absolute value detector, ABS, 406. The absolute value of the auscultation signal is further supplied to the low-pass filter 407 having a cut-off frequency of about 3Hz such that frequency components above 100Hz is diminished. A second branch supplies the auscultation signal to the filter 401 in which the double derivative of the auscultation signal is calculated. The frequency response of the filter 401 corresponds to a second order high-pass slope or an A-weighing approximation in the frequency range below 2000Hz. The double derivative of the auscultation signal is supplied to a low-pass filter 402 in which the relatively high frequency noise components enhanced by the differentiation in the filter 401 is damped. The filter 402 may have a cut-off frequency of about 100Hz. The signal that is low-pass filtered by the filter 402 is supplied to a rectifier or an absolute value detector, ABS, 403. The absolute value of the auscultation signal is further supplied to the low-pass filter 407 that diminishes frequency components above 100Hz.

The decimators, DEC, 405 and 408 generate a reduced sample rate fs_R, that e.g. may be 200Hz since frequency components above 100Hz is diminished. Thus, the computational effort required in the succeeding signal processing of the auscultation signal may be reduced.

The band-pass filter 411 comprising the band-pass filters BPF#1 and BPF#2 receives the decimated auscultation signals and calculates band-pass filtered signals. The band-pass filter may have a pass-band corresponding to the range of expected heart rates e.g. from 0.5 to 3.3HZ.

Note that DC components and frequency components above 25Hz are diminished. The centre frequency of the band-pass filter may be adjusted adaptively in order to enhance the performance of the rhythm estimator. This is discussed later.

The band-pass filtered signals thus represent the envelopes of the auscultation signal and the double differentiated auscultation signal at a reduced sample rate.

The decimators, DEC, 412 and 413 generate a further reduced sample rate that may be e.g. 50Hz since frequency components above 25Hz are diminished. Thus, the computational effort required in the succeeding signal processing of the auscultation signal may be reduced even further.

Samples from the band-pass filter are supplied to the buffers 414 and 415 at even points in time. The buffers may e.g. be circular buffers that have a capacity for a number of sample points, e.g corresponding to a signal sequence of 5 seconds. Samples are taken from the buffers when the estimator 422 is initiated. The estimator may be initiated at certain regular or irregular points in time e.g. in order to provide information for the update of a digital display.

The samples that are taken from the buffers 414 and 415 are scaled by means of the auto-scale means 416 and 417 respectively. The auto-scale means may scale the received sample values by ensuring that the peak value over a number of scaled samples take the value 1.0, e.g. the number of samples may correspond to the signal sequence stored in the buffers 414 and 415, that is the signal sequence is normalised to a peak value equal to 1.0. The scaled samples are added in the adder 418 with equal weights since the information in the two signals are of equal im-

portance, and the result of the addition is provided as a sum-signal. Alternatively, the two signals may be added with different weights.

The conformity of the sum-signal is calculated by means
5   of the correlator, X-Corr, 419. In a preferred embodiment the correlator comprises means for selecting that cycle in the envelope signal that has the largest amplitude. The time extension of the selected cycle is found by searching for the nearest local minima that is less than
10 zero. Thereby, falling into local minima with a magnitude greater than zero is avoided. This search for a minima is applied both forward and backward relative to the position of the largest amplitude. The selected cycle is stored and cross-correlated with the rest of the envelope
15 signal.

The peak-locater 420 identifies extreme values in a signal representative of the result of the cross-correlation. This may e.g. be carried out by calculating the first derivative of the result of the cross-
20 correlation and identifying inflexions. Preferably the peak-locater identifies maximums in the signal representative of the result of the cross-correlation.

The locations of the extreme values are preferably processed in a quality estimator 421. The estimator decides
25 whether the result of the cross-correlation qualifies for estimating the rhythm, if any, of the auscultation signal. The quality estimator 421 may comprise three steps in the process of estimating the quality:

1. The mean of the time differences between two extreme
30   values must be within predetermined limits;

2. The minimum and maximum time differences in proportion to the mean of the time differences must be within predetermined limits;

3. The magnitude of the result of the cross-correlation at the extreme values location must be within predetermined limits.

If one or more of the above mentioned items 1-3 fail, then it is determined that it is not possible to estimate any rhythm in the auscultation signal. Alternatively, if the above mentioned items pass, then it is possible to estimate a rhythm in the auscultation signal. The rhythm may be estimated by counting the number of extreme values in the result of the cross-correlation per time unit and displaying this as a number of beats per minute.

In alternative embodiments the correlators 306 or 419 can be adapted to calculate an auto-correlation function (i.e. two identical signals having a length corresponding to the capacity of the buffers 303 or 414, 415 are correlated with each other). A signal resulting from the calculation of the auto-correlation function can be used to estimate the rhythm of the auscultation signal and to examine the quality of the auscultation signals. A primary frequency of the auscultation signal is found by searching for a local maximum in the autocorrelation signal. The search is started from 0.3 seconds (200 bpm) and the primary frequency is defined as the first local maximum which exceeds a fixed threshold (if no local maximum is found before a 2 second limit in the autocorrelation, no primary frequency is determined). Before the primary frequency is approved, the following items has to be validated:

1. The mean level of the last 2 seconds of the buffer signal has to exceed a fixed threshold. This valida-

tion is done to prevent estimates based on the wrong
kind of signals (or no signal at all).

2. There has to exist a peak in the autocorrelation lo-
cated twice the time as the first peak (± approx
7.5%). This check is done to validate whether the
buffer signal is periodic or not. This peak also has
to exceed a fixed threshold.

3. If N peaks are found to the left of the first peak
(peak1), the mean of the values found at the equally
spaced locations = (peak1)/(N+1) → (N·peak1)/(N+1)
has to be less than a fixed threshold. This check is
done to prevent a low heart rate estimate if the enve-
lope contains a component with a beat frequency equal
to (N+1)·(primary frequency).

4. The estimate has to be stable over a time period. If
the primary frequency is determined to be lower than
40 beats per minute (bmp), the estimate has to be sta-
ble (±30%) 8 times = 1.6 second. If the main frequency
is determined to be higher than or equal to 40 bmp,
the estimate has to be stable (±30%) 4 times = 0.8
second. This check is done to increase the validity of
the estimate.

Other validation criteria may be feasible, but the above
criteria has proved to work well.

Fig. 5 shows an adaptive band-pass filter. This filter
may be used instead of the band-pass filter 411 shown in
fig. 4. The adaptive filter may be used in estimation of
the rhythm in the auscultation signal over a relatively
wide range of frequencies while it is avoided that the
signal-to-noise ratio is destroyed, i.e. it is possible
preserve a relatively narrow pass-band of the band-pass
filter.

The band-pass filter 512 receives two input signals I1 and I2 and provides two band-pass filtered output signals O1 and O2. The band-pass filter 512 thus comprises two band-pass filters BPF#1 and BPF#2. Each of the filters BPF#1 and BPF#2 comprises a pass-band LB with a relatively low centre frequency and a pass-band HB with a relatively high centre frequency. One of the pass-bands may be selected by supplying an enable signal to either the input LB or HB. This pass-band selection is carried out by the adaptive controller 513.

The high-pass filter 501 receives an input signal corresponding to one of the signals input to the band-pass filter 512. The high-pass filter 501 has a cut-off frequency of about 0.3Hz thereby diminishing very low frequency components that is not used in the adaptive pass-band selection. The filter 501 supplies a high-pass filtered signal to the low-pass filter 502. The low-pass filter has a cut-off frequency of about 1Hz.

The buffers 503 and 504 are capable of storing sample values corresponding to a signal sequence with a time duration of about 3 seconds, respectively. The expected or mean value of the sequences stored in the buffers are calculated by means of the mean value detectors 505 and 506. The mean value calculated by mean value detector 505 represents the mean value of the entire signal input to the band-pass filter 512. The mean value calculated by mean value detector 506 represents the mean value of the low-frequency part, i.e. below about 1Hz, of the signal input to the band-pass filter 512.

Thus, the divider 507 calculates the fraction A/B of the signal that is relatively low-frequent. In an initial situation the threshold Th. is equal to the value 0.45. This value is subtracted from the fraction A/B by means

of the adder 508. In the decision state 509 it is veri-
fied whether the frequency ratio A/B is greater than the
threshold value. If this is the case (T) then the thresh-
old value Th. is set equal to the value 0.35 and the
pass-band LB having a relatively low centre frequency is
selected. This is carried out under the assumption that
the auscultation signal originates from a slow beating
heart. Thus it is possible to diminish higher and super-
fluous frequency components by selecting the pass-band LB
having a relatively low centre frequency. Preferably, the
relatively low centre frequency is about 0.7Hz.

Alternatively, if the frequency ratio A/B is less than
the threshold value Th.(F), then the threshold value Th.
is set equal to the value 0.55 and the pass-band HB hav-
ing a relatively high centre frequency is selected. This
is carried out under the assumption that due to the lack
of low-frequency content in the auscultation signal, the
auscultation is likely to originate from a very fast
beating heart, e.g. from a child. By selecting the pass-
band HB having a relatively high centre frequency there
is maximum likelihood that the rhythm of the auscultation
signal is estimated correctly. Preferably, the relatively
high centre frequency is about 1.2Hz.

Fig. 6 shows signals related to an embodiment using a
correlator and a bias filter comprising A-weighing. The
auscultation signal VCG and the corresponding band-pass
filtered envelope of the VCG signal is shown as a func-
tion of time. Further, the A-weighted or double differen-
tiated auscultation signal d^2VCG/dt^2 and the correspond-
ing band-pass filtered envelope is shown. In view of a
graphical interpretation it is clear that the heart rate
could be estimated by using only one of the band-pass
filtered signals. However, a stronger estimate is pro-
vided if the signals are added and used as input to a

correlator as described above. This is shown by means of the signals `XC-I' and `XC-O'.

Fig. 7 shows an example of when the information in the envelope of the auscultation signal is insufficient. The

5   auscultation signal VCG and the corresponding band-pass filtered envelope of the VCG signal is shown as a function of time. Further, the A-weighted or double differentiated auscultation signal d^2VCG/dt^2 and the corresponding band-pass filtered envelope is shown. In view of a

10 graphical interpretation it is clear that the strongest estimate is provided by the double differentiated auscultation signal d^2VCG/dt^2. However, for the vast majority of auscultation signals the signal `XC-O' provides the strongest overall estimate.

15 Fig. 8 shows an example of when the information in the envelope of the A-weighted auscultation signal is insufficient. The auscultation signal VCG and the corresponding band-pass filtered envelope of the VCG signal is shown as a function of time. Further, the A-weighted or

20 double differentiated auscultation signal d^2VCG/dt^2 and the corresponding band-pass filtered envelope is shown. In view of a graphical interpretation it is clear that the band-pass filtered VCG envelope signal provides the strongest estimate. But, again, the signal `XC-O' pro-

25 vides the strongest overall estimate.

The invention may be embodied as a part of a stethoscope or any other instrument or apparatus. Alternatively, the invention may be embodied as a computer program or a part of a computer program, which may be loaded into the mem-

30 ory of a computer and executed therefrom. The computer program may be distributed by means of any data storage or data transmission medium. The storage media can be magnetic tape, optical disc, compact disc (CD or CD-ROM),

mini-disc, hard disk, floppy disk, ferroelectric memory, electrically erasable programmable read only memory (EEPROM), flash memory, EPROM, read only memory (ROM), static random access memory (SRAM), dynamic random access

5   memory (DRAM), ferromagnetic memory, optical storage, charge coupled devices, smart cards, etc. The transmission medium can be a network, e.g. a local area network (LAN), a wide area network (WAN), or any combination thereof, e.g. the Internet. The network may comprise wire

10 and wire-less communication links. Via the network a software embodiment (i.e. a program) of the invention, or a part thereof, may be distributed by transferring a program via the network.